NON-INVASIVE QUANTIFICATION OF TORSIONAL SHEAR STRESSES ALONG THE TIBIA DURING RUNNING

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INTRODUCTION

The tibia has been known to be the most common site of running injuries such as stress fracture. Scientific knowledge about the prevention of tibial running injuries may be obtained from the quantitative investigation of tibial stress distribution while running because repetitive overloading in a certain area of the bone can result in stress fracture. Moreover, if the tibial stresses could be quantified non-invasively, such a methodology may become clinically applicable in future.

To date, shear stress distribution along the tibia loaded with torsion during running has not been quantitatively understood although human long bones have been considered most susceptible to this type of loading. The purpose of this study, therefore, was to quantify non-invasively the distribution of torsional shear stresses along the tibia during running with the finite element method (FEM) accompanied by the three-dimensional (3-D) inverse dynamics. We also investigated how strongly inter-individual variability of the maximum stress depended on loading factor and geometrical one.

METHODS

Five male healthy university students served as subjects. Their mean age, height, and weight was 21.8 ± 1.5yrs, 173.6 ± 5.9cm, 61.6 ± 7.1kg, respectively (mean ± S.D.). Light reflective spherical markers were attached to each subject’s anatomical landmarks of the lower extremity prior to the trials.

A Kistler force platform (Type 9287A, Kistler, Switzerland) was mounted in the middle of a 20m indoor runway. Two high-speed cameras (HSV-500C3, NAC, Japan) were located around the force platform. Two sets of photocells (Speedtrap II, Brower, USA) were also set to measure mean running speeds.

The subjects were asked to run along the runway at 2.0, 3.5, and 5.0m/s. The ground reaction force and kinematic data were synchronized and sampled at 125Hz during the course of each trial. Five successful trials were recorded for each subject at each speed.

The kinematic data were digitized and converted into 3-D spatial coordinates by the direct linear transformation method. The raw data were filtered with a fourth-order, zero-lag Butterworth digital filter with cut-off frequency at 12Hz. The anthropometric data were collected by individual measurements of the subjects. The locations of the center of mass and inertia properties of each body segment were determined based on established data.

Each subject’s right foot and shank were modeled as system of coupled rigid bodies based on the markers attached. To specify the three-dimensional orientation and motion of each segment relative to the global system, The Z-Y-Z Euler angles (θ, ϕ, and ψ) were used.

Tibial torsional moments during the stance phase of running were quantified by the methodology that we have previously reported (Kawamoto et al., 2002a,b). Net intersegmental moments acting at both ends of the tibia were firstly calculated using inverse dynamics with respect to the global system. Secondly, net axial moments acting at both tibial ends were calculated as the products of the direction cosine of the longitudinal axis of the tibia and the net intersegmental moments acting at both tibial ends. Finally, the net axial moments were determined as tibial torsional moment (Mtore) based on the quasi-equilibrium balance assumption in which these axial moments matched well with each other during the stance phase of running.

To develop the 3-D finite element model of each subject’s right tibia, MRI images of the cross-sections of the shank were obtained by every 10mm and then scanned. The inner and outer contour profiles of the tibial cortical bones were digitized. The 3-D shape of the tibia was reconstructed from these cross-sectional data. The tibial models were divided into hexa elements and penta ones using the FEM software (FEMLEEG, HOCTSYSTEMS, Japan, Figure 1). The total number of the elements and that of the nodes were 1036-1132 and 2121-2249, respectively.

Figure1: An example of the finite element model of the tibia used in the present study (Sub. 5).

The boundary and load conditions for the stress analyses were assumed as follows; 1) The tibia was bound at the proximal end of it, and 2) Torsional moment was applied onto the midpoint of the distal end of the tibia. These assumptions were based on the identification of the validity of the quasi-equilibrium balance assumption of the moment.
Mean $M_{tor}$ obtained from five running trials was used for the stress analysis for each subject.

As regards the tibial mechanical property, the cortical bone was assumed to be isotropic. The Young’s modulus and Poisson’s ratio of the bone were 19.3GPa and 0.33, respectively. Only the cortical region was assumed to be loaded.

In the above assumptions, we analyzed the distribution of principal shear stresses ($\tau$) along the whole tibia of every subject during the stance phase of running.

Multiple regression analyses were used to determine how strongly the maximum stress depended on the load (i.e. torsional moment) and on the tibial geometry (i.e. cross sectional area of the tibial cortical region). A significance level of 0.05 was used for the statistical analyses.

RESULTS AND DISCUSSION

Figure 2 shows the individual $M_{tor}$-vs-time curves during the stance phase of running at 3.5m/s. Figure 3 shows the mean (N=5) $M_{tor}$-vs-time curves at different speeds. The tibial torsional moments generally reached the maximum in the later (50-70%) portion of the midstance at any speeds, although considerable inter-individual differences of the $M_{tor}$ pattern as well as of the magnitude could be found (Fig. 2).

These inter-individual differences seemed due to the frontal shank kinematics as the same with our previous report (Kawamoto et al., 2002b). The positive action of the moments suggested that the tibiae tended to be loaded in the internal/external rotational direction of the distal/proximal end during the stance. The $M_{tor}$ patterns agreed well with the patterns of the tibial axial moments reported in the previous study (Bellchamber et al., 2000).

The mean values of the maximum moments were 11.1 ± 4.8, 14.2 ± 8.3, and 18.1 ± 10.1Nm for 2.0, 3.5, and 5.0m/s, respectively (mean ± S.D., Tab. 1). These results suggested that the tibial torsional loading tended to increase linearly as running speed increased within the sub-maximum range.

Figure 2: Individual $M_{tor}$-vs-time curves during the stance phase of running at 3.5m/s.

Figure 3: Mean $M_{tor}$-vs-time curves during the stance phase of running at different speeds.

Figure 4 shows two examples (Sub. 2: stress-concentrated type and Sub. 3: stress-distributed type) of the stress distribution changes by time while a single stance of running.

In most subjects (except for sub.4 whose moment was smaller than those of the other subjects, see Fig. 2), the principal shear stress ($\tau$) reached the maximum in the later portion of the midstance along with the increase of the $M_{tor}$. The stresses tended to concentrate on the side part of the tibia although it depended on individuals on which side (i.e. inside or outside) higher stresses concentrated.

Figure 4: Examples of the stress distribution changes by time during the stance phase of running (Sub. 2 and 3 at 3.5m/s).
Figure 5 shows the individual stress ($\tau$) distributions along the tibiae at the time the $M_{tor}$ reached the maximum during running (3.5m/s). Figure 6 shows the distributions of the mean stresses ($N=5$) at different running speeds. Although inter-individual variability of the stress distribution was considerable (Fig. 5), the higher stresses tended to concentrate around the distal one-third portion of the tibia (Fig. 6). This result agreed with the epidemiological outlook that this site is the most common site of tibial overuse injuries seen in runners (e.g. Krivickas, 1997). Moreover, in two subjects (Sub. 2 and 5), concentration of the high (up to 20MPa) stresses on this site was much more considerable than other three subjects. It may be possible that these subjects have a higher risk for tibial running injuries associated with torsional loading.

The mean values of the maximum stresses were 10.5 ± 4.4, 13.3 ± 7.7, and 17.1 ± 9.3MPa for 2.0, 3.5, and 5.0m/s, respectively (Tab. 1). These results suggested that the maximum stress level tended to increase linearly along with the moment as running speed increase. The stress values, still, were well below the ultimate shear stress of cortical bone (75MPa, Nigg and Herzog, 1999). This result could support the quantitative validity of the stress values in the present study.

As shown in Table 2, the maximum shear stress could be significantly regressed with the torsional moment and with the CSA of the tibia ($R \geq 0.992$, $P<0.001$). The standard regression coefficient (SRC) for the moment was extremely high (i.e. more than 0.998) at every running speed in contrast the SRC for the CSA was negligible (i.e. less than 0.007). These results suggested that the inter-individual variability of the maximum torsional shear stress during running could depend much more strongly on the loading factor than on the geometrical one. Our findings could support the significance of biomechanical quantification of the tibial loading during various human movements from the viewpoint of the prevention of tibial overuse injuries.

Tibial geometry such as the CSA did not affect the maximum level of the shear stress during running. It should be considered, therefore, the tibial geometry could determine the distribution of the stresses rather than the maximum levels of them.

The present study focused on the torsional shear stresses, taking the mechanical property of the tibia into consideration. However, complex effects of various loading (e.g. bending, compression, and tension) should be further analyzed in order to understand comprehensive stress distribution of the tibia in running.

**CONCLUSION**

The present study quantified non-invasively the distribution of the torsional shear stresses along the tibia during running with the finite element method (FEM) accompanied by the 3-D inverse dynamics. Our main findings were as follows; 1) the tibial torsional shear stress reached the maximum (17.1MPa at 5.0m/s) around the distal (20-30%) part of the tibia during the later portion of the midstance (50-70%) of
running. 2) the stress level tended to increase linearly as running speed increase, and 3) inter-individual variability of the maximum shear stress much more strongly on the the loading factor (i.e. torsional moment) than on the tibial geometrical factor (i.e. cross-sectional area).

The present findings will help us to discuss the mechanism of some types of tibial running injures and effective preventions of them.

ACKNOWLEDGEMENT

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REFERENCES


Table 1: Maximum shear stresses, maximum torsional moments and cross sectional areas (CSAs) of the tibiae where the maximum stress could be found.

<table>
<thead>
<tr>
<th></th>
<th>τ^peak [MPa]</th>
<th>M^peak [Nm]</th>
<th>CSA [mm^2]</th>
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<tbody>
<tr>
<td></td>
<td>2.0m/s</td>
<td>3.5m/s</td>
<td>5.0m/s</td>
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<tr>
<td>Sub.1</td>
<td>13.3</td>
<td>19.1</td>
<td>27.9</td>
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<tr>
<td>Sub.2</td>
<td>11.9</td>
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<td>16.1</td>
</tr>
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<td>4.3</td>
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<td>Sub.5</td>
<td>15.2</td>
<td>23.2</td>
<td>25.1</td>
</tr>
<tr>
<td>Average</td>
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<td>SD</td>
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Table 2: Results of the multiple regression analyses for the maximum stresses in running at different speeds.

<table>
<thead>
<tr>
<th>Speed</th>
<th>R</th>
<th>SRC</th>
<th>M^peak [Nm]</th>
<th>CSA [mm^2]</th>
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<td>2.0m/s</td>
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<td>0.999 **</td>
<td>-0.027 ns</td>
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<tr>
<td>3.5m/s</td>
<td>0.996 **</td>
<td>0.998 **</td>
<td>-0.002 ns</td>
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<tr>
<td>5.0m/s</td>
<td>0.997 **</td>
<td>0.998 **</td>
<td>0.007 ns</td>
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</table>

*: P<0.05, **: P<0.01, ns: not significant.
R: Correlation Coefficient.
SRC: Standard Regression Coefficient.